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Impact of lumbar spine posture on thoracic spine motion and muscle activation patterns ☆

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ABSTRACT

Complex motion during standing is typical in daily living and requires movement of both the thoracic and lumbar spine; however, the effects of lumbar spine posture on thoracic spine motion patterns remain unclear. Thirteen males moved to six positions involving different lumbar (neutral and flexed) and thoracic (flexed and twisted) posture combinations. The thoracic spine was partitioned into three segments and the range of motion from each posture was calculated. Electromyographical data were collected from eight muscles bilaterally. Results showed that with a flexed lumbar spine, the lower-thoracic region had 14.83° and 15.61° more flexion than the upper- and mid-thoracic regions, respectively. A flexed lumbar spine significantly reduced the mid-thoracic axial twist angle by 5.21° compared to maximum twist in the mid-thoracic region. Functional differences emerged across muscles, as low back musculature was greatest in maintaining flexed lumbar postures, while thoracic erector spinae and abdominals showed bilateral differences with greater activations to the ipsilateral side. Combined postures have been previously identified as potential injury modulators and bilateral muscle patterns can have an effect on loading pathways. Overall, changes in thoracic motion were modified by lumbar spine posture, highlighting the importance of considering a multi-segmented approach when analyzing trunk motion.

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1. Introduction

With much of spine biomechanics research focused on the lumbar spine, few studies have investigated the interactions between lumbar posture and thoracic movement and the accompanying trunk muscle activation patterns. Considering the spine itself is not a rigid column it has been suggested that a multi-segment approach is necessary to fully understand task-dependent kinematic movement patterns (Preuss & Popovic, 2010). As people tend to move in a variety of postures at home (e.g. bending forward and twisting to empty a washing machine) and at work (e.g. mechanic repairing manufacturing line machinery), it is important to understand the interactions among different regions of the spine during typical postures that involve both flexing and twisting to further develop our knowledge of potential injury mechanisms.

A recent review article by Briggs, Bragge, Smith, Govil, and Straker (2009) focused on the prevalence and factors for developing thoracic spine pain in the adult working population. These authors found an overall high median prevalence rate of about 30% across all occupations, suggesting that thoracic spine pain could be a significant occupational health problem (Briggs et al., 2009). Specifically, occupations such as health professionals and manual laborers reported the highest lifetime (77%), and one year (54.8%) prevalence, respectively (Briggs et al., 2009); and these types of professions often involve standing in awkward, multi-axis postures. It has been suggested that changes in load distribution as a result of coupled flexion and twisting postures are a mechanism of injury in the lumbar spine (Drake & Callaghan, 2008). Much less is known about the injury mechanisms of the thoracic spine compared to the lumbar spine, so it is important have baseline knowledge of the range of motion (ROM) of the thoracic spine during multi-axis movements in order to further develop our understanding of thoracic spine mechanics.

Thoracic spine ROM is often studied from sitting in a neutral starting position with the influence of other postures neglected, both *in vitro* (Busscher et al., 2009; Oxland, Lin, & Panjabi, 1992; Panjabi, Brand, & White, 1976) and *in vivo* (Gregersen & Lucas, 1967; Willems, Jull, & Ng, 1996). For example, complex motion, or movement involving multiple planes, of the thoracic spine was investigated by Edmondston et al. (2007) who found significant decreases in axial twist angle when the upper body measured at T_6 was flexed while sitting. Consistent to both *in vivo* and *in vitro* literature, the amount of axial twist is generally smallest in the lower-thoracic (T_9–T_{12}) regions relative to the mid- (T_4–T_8) and upper- (T_1–T_4) thoracic regions (Gregersen & Lucas, 1967; Pearcy & Tibrewal, 1984; Preuss & Popovic, 2010; White & Panjabi, 1990; Willems et al., 1996); whereas flexion angles tend to increase inferiorly (Preuss & Popovic, 2010; White & Panjabi, 1990; Willems et al., 1996). However, a gap in the literature remains when considering combined motion of the thoracic region from a standing posture in both neutral and non-neutral starting positions.

Furthermore, muscle activation patterns required to maintain complex trunk postures have yet to be investigated concurrently. It is likely that different combinations of lumbar and thoracic postures (neutral, flexed, or twisted) will alter the patterns of muscle activity, and changes in muscle activation contribute to altering joint level loading which affects risk of injury. It is important to consider muscle activations from different levels of the trunk when considering the motion of the whole spine as the functions of each area may differ depending on the task demands.

Therefore, the purpose of this study was to quantify differences in standing flexion and axial twist ROM of the thoracic spine from neutral and non-neutral lumbar starting positions, and the corresponding muscle activations required to hold each posture. We hypothesized that thoracic flexion would increase in a superior–inferior direction and an increased range of thoracic flexion would occur with increased lumbar flexion. We also hypothesized that axial twist of the mid-thoracic region would be greatest in a neutral starting position yet would decrease during combined flexion and twisting. These hypotheses were based from the results of seated thoracic motion by Willems et al. (1996) and Edmondston et al. (2007). The analyses of the muscle activation patterns were intended to compliment the kinematic results so no formal hypotheses were formulated; however, we generally expected to see bilateral differences during the off-axis movements as well as greater activations from the non-neutral starting postures.

2. Methods

2.1. Participants

Thirteen right-hand dominant males free of neck, back, and shoulder pain for at least one year prior to collection were recruited by convenience sampling from the University population. A male-only population was used due to the required attire to adhere the reflective markers (See Section 2.2. below). Mean ± *SD* age, height, and weight were 23.6 ± 2.1 y, 1.82 ± 0.08 m, and 84.3 ± 13.9 kg, respectively. Informed consent was obtained prior to any measures being taken and all protocols were approved by the Office of Research Ethics at York University.

2.2. Instrumentation

Kinematic data were recorded at a sample rate of 50 Hz (Vicon MX, Vicon Systems Ltd., Oxford, UK) from 63 passive-reflective markers adhered to the skin over select landmarks. Specifically for the spine, this included 24 markers being taped on eight rigid plates (three markers per plate) with the plates being adhered over select spinous processes (T_1, T_4, T_5, T_8, T_9, T_{12}, L_1, and the posterior-superior iliac spines (PSISs)) using double-sided tape. The marker–plate and plate–skin interfaces were inspected prior to and monitored during collection to ensure continuous adherence. The spinous processes were located by palpation of the C_7 and counted downward to L_1. Manual identification of landmarks has been used previously in studies of thoracic motion using surface-based marker techniques (Edmondston et al., 2007; Nairn, Chisholm, & Drake, 2013; Preuss & Popovic, 2010; Willems et al., 1996). The remaining markers were placed on the upper legs, pelvis, upper arms, and head. The reflective markers on the plates adhered to the spinous processes were used to define four rigid segments: upper-thoracic (T_1–T_4), mid-thoracic (T_5–T_8), lower-thoracic (T_9–T_{12}), and the lumbar segment (L_1–PSISs) (Nairn et al., 2013).

Electromyographical signals were collected bilaterally from eight muscles: *external oblique* (15 cm lateral to the umbilicus at a 45° angle (Marras & Mirka, 1993; McGill, 1991)); *internal oblique* (below *external oblique* and approximately midway between the anterior superior iliac spine and symphysis pubis, above the inguinal ligament (Cholewicki & McGill, 1996)); *rectus abdominis* (3 cm lateral to midline of abdomen, 2 cm above umbilicus (Drake, Fischer, Brown, & Callaghan, 2006; Marras & Mirka, 1993)); *latissimus dorsi* (most lateral portion of the muscle at the T_9 level (Drake et al., 2006; McGill, 1991)); *upper-thoracic erector spinae* (*ES*) (largest muscle mass approximately 2.5 cm lateral to T_4 spinous process (Burnett et al., 2008)); *lower-thoracic ES* (largest muscle mass approximately 4 cm lateral to T_9 spinous process (Drake et al., 2006; McGill, 1991)); *lumbar ES* (largest muscle mass approximately 4 cm lateral to L_3 spinous process (Drake et al., 2006; McGill, 1991)); and the superficial fibers of lumbar multifidus (*multifidus*: at L_5 parallel to a line connecting the PSIS and L_1–L_2 interspinous space (Dankaerts, O'Sullivan, Burnett, & Straker, 2006)). The signals were differentially amplified (frequency response 10–1000 Hz, common mode rejection 115 dB at 60 Hz, input impedance 10 GΩ; two of model AMT-8, Bortec, Calgary, Canada), and converted from an analog to digital signal at a rate of 2400 Hz (Vicon MX motion capture system, Vicon Systems Ltd., Oxford, UK). The skin was shaved then swabbed with alcohol swab to help with adherence of the electrodes. The electrodes were disposable silver/silver-chloride surface EMG electrodes with a center-to-center spacing of 2.5 cm (Ambu® Blue Sensor N, Ambu A/S, Denmark). All trials were video recorded using a Sony Handycam® (Model HDR-CX110) and shown through a live feed to a television screen.

2.3. Data collection

After consent and EMG preparation, a five minute EMG rest trial was collected with the participant lying supine on a therapy table. This was followed by the maximum voluntary contraction (MVC) procedures: the modified back extension (McGill, 1991), sit-up (McGill, 1991), and lateral pull-down (Drake et al., 2006), which were designed to elicit maximal activations from each muscle for normalization purposes. Each MVC posture was repeated three times with a minimum three minutes rest between each trial to minimize fatigue effects.

After the MVC protocol the reflective markers were applied to the participant as outlined above. Participants then performed three consecutive repeats of six standing postures presented in a random order: maximum forward flexion (*Max Flexion*), maximum right axial twist (*Max Twist*), and four postures that were combinations of a neutral or flexed lumbar region, to approximately 50% of *Max Flexion*, with a flexed or right twisted thoracic region. Specifically, the combination postures were: *Lumbar-neutral–Thoracic-flexed*, *Lumbar-flexed–Thoracic-flexed*, *Lumbar-neutral–Thoracic twisted*, and *Lumbar-flexed–Thoracic-twisted*). These postures were chosen to represent combinations that commonly occur in real-life either at work and/or during activities of daily living. For the *Max Flexion* and *Max Twist* trials, participants were instructed to bend/rotate as far as possible without specific instruction on how to move the different spine regions. For the *Lumbar-flexed* trials an acetate sheet was placed over the television screen and was used to mark the upright, maximum lumbar flexion, and approximately 50% of lumbar flexion prior to the completion of the posture trials. This allowed for an estimation of 50% lumbar flexion and ensured each participant started from an upright neutral position.

Participants were instructed to keep their knees straight and their "hip bones" (anterior–superior iliac spines) pointed forward during all trials except in *Max Twist* (where they were instructed to twist maximally). Participants were also told to let their arms hang naturally during the – *thoracic-flexed* and *Max Flexion* trials and to follow the trunk motion during the – *thoracic-twisted* and *Max Twist* trials (Fig. 1). During the *Thoracic-flexed* trials participants were instructed to curl their chin to their chest after achieving the starting lumbar posture of either neutral or flexed. For all trials, when participants reached their maximum angle, the posture was held for 5 s before returning to upright. Fig. 1A and B depicts one participant performing the *Lumbar-neutral–Thoracic-twisted* and the *Lumbar-flexed–Thoracic twisted* trials, respectively, and 1c shows the rigid segments defined by the markers.

2.4. Data processing

All kinematic and EMG data were processed using Visual3D v4.0 (C-Motion Inc., Germantown, USA). The kinematic data were filtered using a dual-pass 4th order Butterworth low-pass filter with

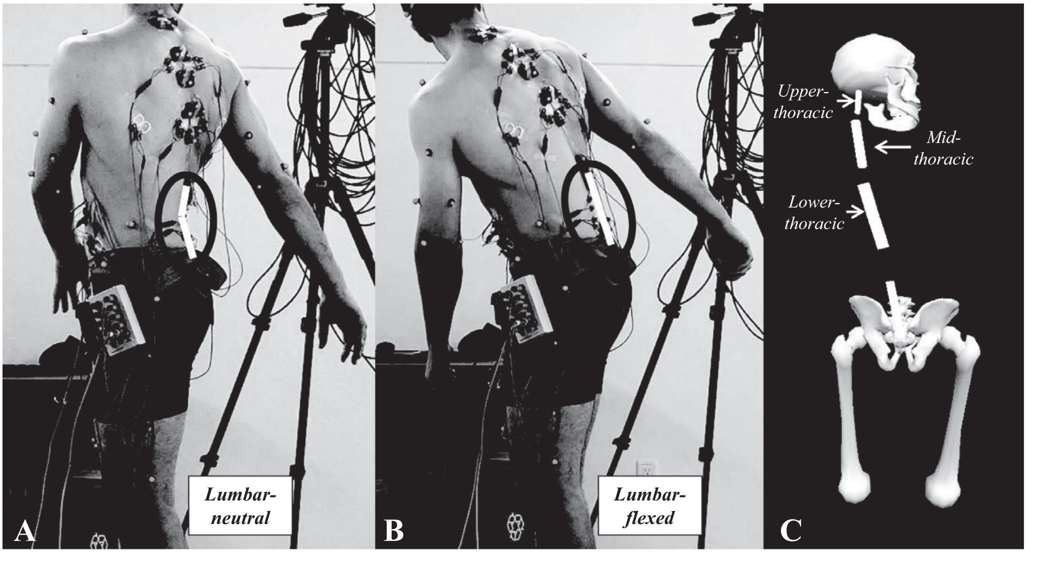

Fig. 1. (A) Example of the *Lumbar-neutral–Thoracic-twisted* posture. The neutral lumbar position is highlighted by the circled area indicating upright lordosis. (B) Example of the *Lumbar-flexed–Thoracic-twisted* posture. The flexed lumbar position is highlighted by the circled area indicating a mid-range of lumbar flexion. (C) Posterior view of a *Thoracic-twisted* trial illustrating the thoracic segments created from the rigid plates.

a cut-off frequency of 2.5 Hz as determined from residual analysis (Winter, 2009). The creation of each spine segment was based on a proximal and distal plate, using the bilateral markers for segment definition. The markers on the left side of each plate were arbitrarily defined as the medial side while markers on the right side of each plate were designated as the lateral side in order to assist with segment definition. The segment endpoints were computed as the mid-point between the medial and lateral marker location on each plate, with the positive axial (z-) axis defined as the line passing from the distal to proximal endpoints. The frontal x–z plane was computed within the software using a least squares method, followed by determination of the y-axis perpendicular to the frontal plane and z-axis. The x-axis was then calculated perpendicular to the y–z plane according to the right-hand rule. The additional markers positioned over the midline were used for tracking along with the bilateral markers.

Three-dimensional rotational angles were calculated for thoracic angles as the upper segment with respect to the segment below it using an x–y–z (Flexion–Lateral Bend–Axial Twist) Cardan sequence (Nairn et al., 2013). For example, the *upper-thoracic* angle was defined as the rotation of the upper-thoracic segment relative to the mid-thoracic segment and so on for each angle. These regional relative angles were defined as the *upper-thoracic*, *mid-thoracic*, and *lower-thoracic* angles. The lumbar angle used to determine maximum flexion and 50% of flexion was defined as the angle between the plate at L_1 and the plate over the PSISs (Fig. 1).

Raw EMG data were initially passed through a Butterworth high-pass filter with a 30 Hz cut-off to remove heart rate contamination (Drake & Callaghan, 2006), followed by full-wave rectification then low-pass filtered using a 4th order Butterworth filter with a 6 Hz cut-off to produce the linear envelope of the signal. The rest trial was visually inspected and a 30 s portion was selected to average based on the absence of any spikes/noise across all 16 channels. The averaged rest bias from each channel was then subtracted from all subsequent trials followed by obtaining the maximum value for each channel from the MVC trials. Each EMG channel was normalized to its maximum, resulting in the values being reported as a percentage of maximum voluntary contraction (%MVC). The peak EMG values from each trial were found from the hold phase of the trial. This hold phase was determined as the time between when the last segment stopped moving into the posture and when the first segment started moving out of the posture while returning to upright.

Following data processing, a one-way within subject ANOVA with Tukey's HSD post hoc and $\alpha = 0.05$ was run to analyze the percent angle of lumbar flexion during the trials, as the target was mid-range lumbar flexion to approximately 50%Max during the *Lumbar-flexed* trials. The Mean ± *SD lumbar* angle as a percent of maximum flexion for both *Lumbar-flexed–Thoracic-flexed* (33.06 ± 20.83%Max Flexion) and *Lumbar-flexed–Thoracic-twisted* (44.37 ± 13.50%Max Flexion) were less than the 50% target; however, these angles were not significantly different from each other, and were both significantly greater than the *lumbar* angles in the neutral trials, with *Lumbar-neutral–Thoracic-flexed* being 15.92 ± 12.87%Max Flexion, and *Lumbar-neutral–Thoracic-twisted* being 16.86 ± 13.88%Max Flexion. Therefore, the original analyses proceeded as planned and are outlined below.

2.5. Statistical analysis

In order to quantify differences in regional angles across the postures, the flexion and axial twist ROM angles were analyzed using separate 3×6 within subject ANOVAs with the factors being: Region (*upper-thoracic*, *mid-thoracic*, *lower-thoracic*), and Posture (*Lumbar-neutral–Thoracic-flexed*, *Lumbar-flexed–Thoracic-flexed*, *Lumbar-neutral–Thoracic twisted*, *Lumbar-flexed–Thoracic-twisted*, *Max Flexion*, *Max Twist*) using a statistical software package (JMP10, SAS Institute Inc., Cary, NC). Maximum EMG levels (%MVC) from each muscle during the hold phase were analyzed individually using 2×6 within subject ANOVAs with the factors being: Side (*left* or *right*), and Posture (*Lumbar-neutral–Thoracic-flexed*, *Lumbar-flexed–Thoracic-flexed*, *Lumbar-neutral–Thoracic twisted*, *Lumbar-flexed–Thoracic-twisted*, *Max Flexion*, *Max Twist*). For all statistical analyses $\alpha = 0.05$ and significant findings were further analyzed with Tukey's HSD post hoc.

The reliability of the 20 kinematic measures was assessed with intraclass correlation coefficients (ICCs) calculated using a single measure 2-way mixed-model, and evaluated using the thresholds reported in Lee and Granata (2008) and Fleiss (1986). Excellent reliability (>0.750) was shown for 19/20 of the measures (range: ICC(3,1) = 0.771–0.942) with one measure having moderate reliability (0.400–0.749). Specifically, five ICCs were greater than 0.900, 11 were between 0.800 and 0.900, three were between 0.750 and 0.800, and one ICC was 0.492 (*mid-thoracic* axial twist during the *Lumbar-flexed–Thoracic-twisted* trial).

3. Results

Thoracic flexion increased in a superior–inferior direction only during the *maximum flexion* trials, and the *lower-thoracic* region was the only one to further increase flexion angle with increased lumbar flexion ($F_{10,120} = 27.76$, $p < .0001$). Mean ± *SD* flexion ROM during *Max Flexion* was significantly different between all thoracic regions, with *lower-thoracic*, *mid-thoracic*, and *upper-thoracic* values being 35.34° ± 7.83, 11.76° ± 6.13, and 4.29° ± 3.21, respectively (Fig. 2). During the *Lumbar-flexed–Thoracic-flexed* trial the *lower-thoracic* angle (24.91° ± 8.22) was significantly greater than both the *mid-thoracic* (10.08° ± 5.07) and *upper-thoracic* (9.30° ± 4.15) angles (Fig. 2). No clear pattern emerged during the *Lumbar-neutral–Thoracic-flexed* trial as the only significant difference was between the *lower-thoracic* and *upper-thoracic* regions (Fig. 2). Only the *lower-thoracic* angle showed a continuous increase in flexion with increasing lumbar flexion as both the *upper-thoracic* and *mid-thoracic* regions were not significant between each of the postures (Fig. 2).

Mid-thoracic axial twist significantly decreased by 5.21° from the *Max Twist* trial to the *Lumbar-flexed–Thoracic-twisted* trial as shown in Fig. 3 ($F_{10,120} = 4.12$, $p < .0001$). However, there were no statistical differences between the *Lumbar-flexed* – (10.01° ± 3.35) and *Lumbar-neutral* – (14.98° ± 6.20) *Thoracic-twisted* trials in the *mid-thoracic* region (Fig. 3). Between thoracic regions, the twist ROM of the *lower-thoracic* region during the *Lumbar-flexed* (7.16° ± 5.48) and *Lumbar-neutral* (6.65° ± 4.57) trials were significantly less than the *mid-thoracic* region during the *Lumbar-neutral–Thoracic-twisted* and *Max Twist* (15.21° ± 5.18) trials (Fig. 3). None of the *Max Twist* angles were significantly different between regions, with the *upper-thoracic*, *mid-thoracic*, and *lower-thoracic* ROM being 11.82° ± 6.70, 15.21° ± 5.18, and 11.21° ± 5.50, respectively (Fig. 3).

Lumbar ES and *multifidus* did not show bilateral differences, and these muscles showed the greatest activations during the mid-range *Lumbar-flexed* trials (Table 1) ($F_{5,60} > 3.98$, $p < .004$). All other muscles except *rectus abdominis* showed bilateral differences, with the *right* side showing the greatest activations during the *Thoracic-twisted* trials ($F_{5,60} > 5.37$, $p < .0004$). Specifically, *right upper-thoracic*

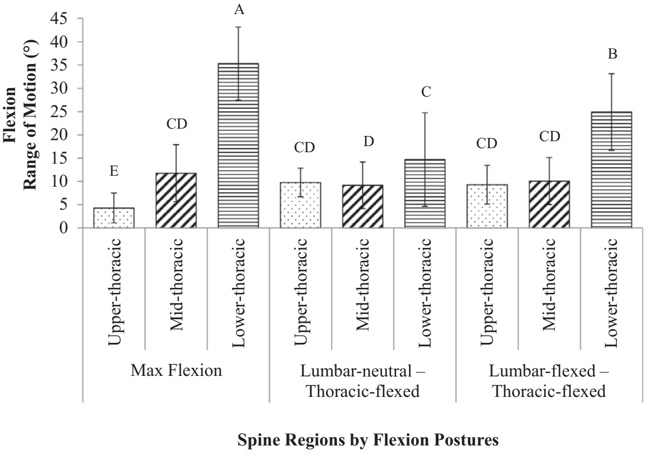

Fig. 2. Mean ± *SD* of the regional flexion range of motion (°) for the flexion-based trials. Postures include: *Max Flexion*, *Lumbar-neutral–Thoracic-flexed*, and *Lumbar-flexed–Thoracic-flexed*. Levels not connected by the same letter are significantly different. α = 0.05 for all comparisons.

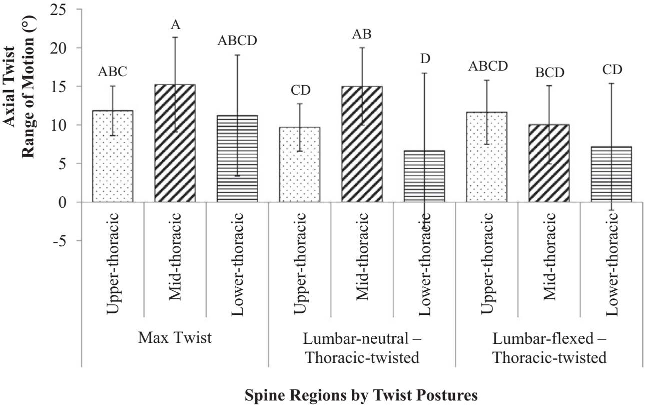

Fig. 3. Mean ± *SD* of the regional twist range of motion (°) for the twist-based trials. Postures include: *Max Twist*, *Lumbar-neutral–Thoracic-twisted*, and *Lumbar-flexed–Thoracic-twisted*. Levels not connected by the same letter are significantly different. $\alpha = 0.05$ for all comparisons.

Table 1
Mean ± *SD* of the peak EMG (%MVC) values from the hold phase of each posture for each muscle.

Lumbar ES	Collapsed right and left		Multifidus	Collapsed right and left	
L_{neut}–T_{flex}	8.80 ± 5.70^{AB}		L_{neut}–T_{flex}	12.25 ± 8.29^{ABC}	
L_{flex}–T_{flex}	11.02 ± 7.14^{A}		L_{flex}–T_{flex}	15.25 ± 5.89^{AB}	
L_{neut}–T_{twist}	4.78 ± 2.57^{B}		L_{neut}–T_{twist}	7.37 ± 6.78^{C}	
L_{flex}–T_{twist}	11.47 ± 5.81^{A}		L_{flex}–T_{twist}	16.69 ± 8.04^{A}	
Max Flex	7.39 ± 8.37^{AB}		Max Flex	10.51 ± 8.60^{BC}	
Max Twist	7.28 ± 5.62^{AB}		Max Twist	10.37 ± 9.31^{BC}	
Upper-thoracic ES	Right	Left	Lower-thoracic ES	Right	Left
L_{neut}–T_{flex}	11.38 ± 0.80^{B}	9.92 ± 8.87^{B}	L_{neut}–T_{flex}	5.76 ± 5.2^{B}	6.80 ± 4.54^{B}
L_{flex}–T_{flex}	8.68 ± 6.85^{B}	11.07 ± 9.42^{B}	L_{flex}–T_{flex}	6.04 ± 5.33^{B}	6.44 ± 4.70^{B}
L_{neut}–T_{twist}	37.63 ± 24.16^{A}	13.77 ± 17.92^{B}	L_{neut}–T_{twist}	17.34 ± 12.90^{A}	7.36 ± 8.55^{B}
L_{flex}–T_{twist}	44.08 ± 26.61^{A}	10.79 ± 7.53^{B}	L_{flex}–T_{twist}	17.37 ± 14.33^{A}	5.31 ± 4.57^{B}
Max Flex	3.07 ± 2.08^{B}	3.47 ± 2.98^{B}	Max Flex	3.45 ± 4.09^{B}	5.12 ± 9.24^{B}
Max Twist	45.98 ± 31.03^{A}	12.57 ± 16.79^{B}	Max Twist	18.49 ± 13.63^{A}	5.92 ± 5.99^{B}
External oblique	Right	Left	Internal oblique	Right	Left
L_{neut}–T_{flex}	5.29 ± 4.55^{EF}	5.75 ± 4.11^{EF}	L_{neut}–T_{flex}	14.15 ± 17.57^{B}	10.74 ± 9.52^{B}
L_{flex}–T_{flex}	5.98 ± 5.94^{DEF}	3.82 ± 3.99^{F}	L_{flex}–T_{flex}	6.54 ± 3.95^{B}	7.48 ± 7.37^{B}
L_{neut}–T_{twist}	17.39 ± 11.85^{BC}	10.20 ± 6.52^{CDEF}	L_{neut}–T_{twist}	37.87 ± 35.13^{A}	12.88 ± 9.53^{B}
L_{flex}–T_{twist}	15.23 ± 9.68^{BCDE}	21.12 ± 14.42^{AB}	L_{flex}–T_{twist}	33.80 ± 25.11^{A}	10.74 ± 7.94^{B}
Max Flex	9.81 ± 5.93^{CDEF}	11.10 ± 8.18^{BCDEF}	Max Flex	13.51 ± 11.86^{B}	15.83 ± 18.77^{B}
Max Twist	30.41 ± 23.73^{A}	16.30 ± 9.14^{BCD}	Max Twist	47.56 ± 33.17^{A}	15.52 ± 11.43^{B}
Rectus abdominis	Right	Left	Latissimus dorsi	Right	Left
L_{neut}–T_{flex}	11.19 ± 15.83	15.57 ± 14.79	L_{neut}–T_{flex}	5.45 ± 8.41^{B}	5.21 ± 5.27^{B}
L_{flex}–T_{flex}	5.96 ± 4.39	12.76 ± 11.35	L_{flex}–T_{flex}	6.79 ± 4.32^{B}	7.63 ± 7.03^{B}
L_{neut}–T_{twist}	7.19 ± 3.94	10.48 ± 6.47	L_{neut}–T_{twist}	12.69 ± 14.35^{AB}	5.87 ± 4.66^{B}
L_{flex}–T_{twist}	10.38 ± 13.28	12.00 ± 7.72	L_{flex}–T_{twist}	22.41 ± 15.78^{A}	7.22 ± 5.07^{B}
Max Flex	16.69 ± 15.20	21.94 ± 25.96	Max Flex	4.75 ± 4.21^{B}	7.95 ± 7.43^{B}
Max Twist	9.25 ± 6.79	10.32 ± 7.26	Max Twist	18.99 ± 20.09^{A}	6.86 ± 4.07^{B}

Note. ES = erector spinae; in each muscle, values not connected by the same letter are significantly different at $\alpha = 0.05$. For example, in latissimus dorsi each posture on the left side is different from the L_{flex}–T_{twist} and Max Twist on the right side. Collapsed muscle groups showed no effects of side. Postures include: *Lumbar-neutral–Thoracic-flexed* (L_{neut}–T_{flex}), *Lumbar-flexed–Thoracic-flexed* (L_{flex}–T_{flex}), *Lumbar-neutral–Thoracic twisted* (L_{neut}–T_{twist}), *Lumbar-flexed–Thoracic-twisted* (L_{flex}–T_{twist}), *Maximum Flexion* (Max Flex), and *Maximum Axial Twist* (Max Twist).

ES, *lower-thoracic ES*, and *internal oblique* showed greater activations in the *Max Twist*, *Lumbar-flexed–Thoracic-twisted* and *Lumbar-neutral–Thoracic-twisted* trials compared to all other trials (Table 1). Additionally, *right latissimus dorsi* showed greater activations during *Max Twist* and *Lumbar-flexed–Thoracic-twisted* (Table 1). *External oblique* showed the most varying patterns, where the *right* side during *Max Twist* was greater than all postures except for *left* side *Lumbar-flexed–Thoracic-twisted* trial (Table 1). Furthermore, *left external oblique* in the *Lumbar-flexed–Thoracic twisted* trial was greater than the *right* side of the same posture (Table 1).

4. Discussion

This study showed that during standing, interactions among different thoracic spine regions emerged when different lumbar postures were adopted. This agrees with Preuss and Popovic (2010) who found that as spine segmentation increases, so too does the interaction complexity and variability of the movement in the absence of pathology during task-dependent movement. In their study, participants were seated and were instructed to lean and touch a target with their head located at various degrees around them. In the current study, the task was not goal-oriented per se; however maximal thoracic range was the end-result of the task. Given the interactions found to influence thoracic ROM, one conclusion is that task-dependent motion was present in a standing posture, and subsequently during movement that could be considered common in daily activities.

The hypothesized superior–inferior flexion pattern was confirmed during the *Max Flexion* trial, similar to the results of Willems et al. (1996) during seated thoracic flexion. There were differences in the numerical values of *Max Flexion* between the results of Willems et al. (1996) and the present study in the *upper-thoracic* and *lower-thoracic* regions. Using electromagnetic sensors, Willems et al. (1996) reported flexion of 8.6° ± 5.0 and 12.7° ± 3.4 in the upper-thoracic, from T_1 to T_4, and lower-thoracic, from T_8 to T_{12}, regions respectively, compared to the respective regional ROM of 4.29° ± 3.21 and 35.34° ± 7.83 in the present study. One reason for the discrepancy likely stems from the postures themselves, as Willems et al. (1996) used a seated posture with a lumbar backrest and specific instructions to curl the trunk forward with minimal motion about the lumbar spine. During standing, full-forward flexion involves greater motion about the lumbar spine rather than the thoracic spine, and flexing both the thoracic and lumbar spine maximally simultaneously while standing is not likely a typical posture that an average person is to adopt day-to-day. When the focus of the movement was about the thoracic spine directly, the ROM of the present study became more similar to those of Willems et al. (1996) For example, during the *Lumbar-neutral–Thoracic-flexed* trial *upper-thoracic* flexion increased to 9.76° ± 3.07 and *lower-thoracic* flexion decreased to 14.71° ± 10.06. Additionally, Preuss and Popovic (2010) found the greatest amount of flexion inferiorly during each task-dependent sitting tasks involving sagittal plane motion with varying degrees of lumbar flexion. It is likely that with participants sitting, the added support and subsequent balance allowed them to progressively flex forward, compared to standing postures where greater balance would be required as the trunk moves over the base of support. These highlight some of the functional differences in movement patterns between sitting and standing postures, and further exemplify the interactions between the lumbar and thoracic regions.

The reduction of *mid-thoracic* twist ROM during a flexed posture agreed with the results of Edmondston et al. (2007), who found significant decreases in axial twist angle (measured at T_6) when the thorax was flexed. Additionally, Montgomery, Boocock, and Hing (2011) found increases in whole-trunk rotation when the trunk was flexed and the spine remained neutral, yet found decreases in rotation when the spine itself was flexed. This remains consistent with the current results of a neutral spine showing greater rotation than a flexed spine. Likewise, Preuss and Popovic (2010) found the mid-lower thoracic region from T_6 to T_9 to show the greatest axial rotation in each twisting task with a progressive decrease at the more caudal and rostral levels. Edmondston et al. (2007) used a rotation angle relative to the laboratory space, so the values are not directly comparable; however, the axial twist angles reported by Willems et al. (1996) were similar to the present study. During seated maximum twist Willems et al. (1996) reported the *upper-thoracic*, *mid-thoracic*, and *lower-thoracic* angles to be 13.6° ± 5.0, 23.3° ± 6.1, and 9.1° ± 5.4, respectively. These were slightly less in value, yet comparable in pattern to the *Lumbar-neutral–Thoracic-twisted* posture in the present study, which found *upper-*, *mid-*, and *lower-thoracic* ROM to be 9.68° ± 6.39, 14.98° ± 6.20, and 6.65° ± 4.57, respectively. During *Max Twist* the values were also comparable in the present standing posture to the seated posture of Willems et al. (1996), (Fig. 3) indicating that the change from standing to sitting in thoracic motion may have a greater effect on flexion ROM than it does twist on ROM.

Movement differences between the spine regions can be partly explained by vertebral anatomy. In general, in the absence of flexion the facet orientation of the upper- and mid-thoracic spine (T_1–T_8) facilitates axial twist as compared to general orientation in the lower-thoracic (T_9–T_{12}), where the transition typically occurs between thoracic and lumbar spine (Davis, 1959; Maiman & Pintar, 1992; White & Panjabi, 1990). Additionally, the ribcage provides some limitation to the motion within the thoracic region (Watkins et al., 2005). From an evolutionary perspective (Williams, 2011), the vital organs are contained within the thorax and as humans became bipedal, a need arose to limit the potential for excess motion (e.g. reducing the internal compressive forces) to help protect these key structures (Tortora, 2005). *In vitro* work has shown that axial twist combined with flexion may be an injury modulator in the lumbar spine (Drake, Aultman, McGill, & Callaghan, 2005; Drake & Callaghan, 2009). It remains unclear if this link exists within the thoracic spine; however, it is possible the general reduction of twist angle with flexion in the *mid-thoracic* region, as a result of the anatomical properties such as facet orientation and the role of the ribcage, could be an inherent protective mechanism against unwanted complex motion, thereby reducing the risk of injury.

In terms of potential injury mechanisms in the thoracic spine, there are several factors which could contribute strictly from a postural perspective. Using an optimization model, Briggs et al. (2007) estimated the spinal loading in 21 high-kyphosis and 23 low-kyphosis elderly participants during upright standing. They found that gravitational effects alone on a high-kyphosis spine showed greater normalized shear muscle forces in the upper (T_2–T_5), and lower (T_{10}–L_1) regions (Briggs et al., 2007). Recently, a model from Lee, Lee, and Hayes (2013) estimated the compression forces in the lower thoracic and lumbar regions from intradiscal pressure and cross-sectional area measurements in various upright postures, with and without a 20 kg mass. Highest thoracic compression forces were found while holding the mass in an upright posture with the elbows at 90°, with a 40% decrease in compression when the trunk was flexed and the arms extended downward, which was attributed to a change in external moment arm (Lee et al., 2013). Although the compression force was reduced, there would still be a corresponding increase in the shear forces as a result of the posture change from upright to flexion. In the present study, a flexed and twisted posture would also alter the external moment arm, which would increase the anterior–posterior and lateral shear forces, thereby increasing the overall loading on the thoracic spine.

Functional patterns emerged in the muscle activations at various spine regions depending on the posture. The distal spine muscles of *lumbar ES* and *multifidus* showed peak activation during the *Lumbar-flexed* trials (Table 1) indicating an extensor role needed to maintain the mid-range flexion postures. In contrast the bilateral differences found in the remaining muscles during the twisting trials indicated a functional transition across the ascending levels of the spine, as the primary function changes from an extensor role to a twist role. A possible disconcerting finding was the high activation of the *right upper-thoracic ES* muscle during twisting trials (Table 1). Kumar (2010) measured EMG from the abdominals, lower-thoracic ES, lumbar ES, and latissimus dorsi, and noted axial rotation decreased activations of the ipsilateral external oblique, and contralateral internal oblique and latissimus dorsi, which may result in increased asymmetrical loading on the spine. As industrial lowering tasks often involve rotation (Kumar, 2010), increased ipsilateral *upper-thoracic ES* muscles during twisting could increase the risk of injury, especially if reaching is involved. Different levels of trunk EMG should be considered depending on the task being analyzed, especially in the upper-thoracic region during standing-based twisting. Many functional tasks include this type of movement; however, *upper-thoracic ES* is often not quantified.

One of the issues of using surface mounted markers is the potential for relative skin motion to occur. Care was taken to ensure the plates remained adhered to the skin and any trials where excess motion was observed or suspected as a result of loosened taped were re-done. It is also possible that factors such as muscle contractions may have affected the motion of the plates, as the plates themselves required adherence across the midline of the spine and slightly over the muscle bellies.

Mounting the markers to a rigid plate eliminated any potential relative motion that would have occurred among the individual markers mounted directly onto the skin. Preuss and Popovic (2010) have discussed the issues of skin markers for a multi-segmented approach to spine analysis and ultimately concluded skin mounted motion capture remains the most viable approach for studying the trunk and spine compared to alternatives such as percutaneous screws and medical imaging techniques. Another possible limitation to the study was that pelvis motion was not constrained. With the instrumentation involved, it was not possible to attach an external device to the pelvis without compromising the pelvic markers and EMG channels. Further, the focus of the study was free, voluntary motion, not altered or facilitated motion that would have been generated with the use of a constraint system. Additionally the use of a relatively small sample size of young, male-only participants may limit the generalizability of results across all populations such as clinical groups or the elderly. Regardless, similar sample sizes and population groups have been used in previous studies of trunk kinematic (Claus, Hides, Moseley, and Hodges (2009b) ($n = 10$) and EMG (Claus, Hides, Moseley, & Hodges, 2009a) ($n = 14$) measures. Also, it is necessary to gather information on a non-clinical population in order to establish a baseline for comparison to other groups in the future.

5. Conclusion

To summarize, complex motion is typical both at work and home and knowing how different regions of the spine react to various postures is important in understanding the potential injury mechanisms that can occur as a result of combined flexed and twisted postures. Increases in lumbar flexion tended to decrease axial rotation within the *mid-thoracic* region, which overall had the largest amount of regional twist ROM. Flexion ROM followed a superior–inferior pattern during *Max Flexion* only, whereas increased changes in lumbar starting posture had an increasing effect on *lower-thoracic* flexion, with minimal effects on the *upper-* and *mid-thoracic* areas. Muscle activation patterns showed functional patterns as the lumbar muscles required the largest activations to maintain the mid-range lumbar postures. Twisting resulted in higher activations of ipsilateral *thoracic ES* and *oblique* muscles, which could change the internal loading pathways of these muscles if externally loaded, thereby potentially increasing the risk for injury in a workplace. These results highlight the importance of utilizing a multi-segment approach to analyze spine kinematics to help elucidate the complex interactions associated with non-neutral starting positions and off-axis postures.

Conflict of interest disclosure

None.

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References

Briggs, A. M., Bragge, P., Smith, A. J., Govil, D., & Straker, L. M. (2009). Prevalence and associated factors for thoracic spine pain in the adult working population: A literature review. *Journal of Occupational Health, 51*, 177–192.

Briggs, A. M., van Dieën, J. H., Wrigley, T. V., Greig, A. M., Phillips, B., Lo, S. K., et al (2007). Thoracic kyphosis affects spinal loads and trunk muscle force. *Physical Therapy, 87*, 595–607.

Burnett, A., O'Sullivan, P., Ankarberg, L., Gooding, M., Nelis, R., Offermann, F., et al (2008). Lower lumbar spine axial rotation is reduced in end-range sagittal postures when compared to a neutral spine posture. *Manual Therapy, 13*, 300–306.

Busscher, I., van Dieën, J. H., Kingma, I., van der Veen, A. J., Verkerke, G. J., & Veldhuizen, A. G. (2009). Biomechanical characteristics of different regions of the human spine: An in vitro study on multilevel spinal segments. *Spine, 34*, 2858–2864.

Cholewicki, J., & McGill, S. M. (1996). Mechanical stability of the *in vivo* lumbar spine: Implications for injury and chronic low back pain. *Clinical Biomechanics, 11*, 1–15.

Claus, A. P., Hides, J. A., Moseley, G. L., & Hodges, P. W. (2009a). Different ways to balance the spine: Subtle changes in sagittal spinal curves affect regional muscle activity. *Spine, 34*, E208–E214.
Claus, A. P., Hides, J. A., Moseley, G. L., & Hodges, P. W. (2009b). Is 'ideal' sitting posture real? Measurement of spinal curves in four sitting postures. *Manual Therapy, 14*, 404–408.
Dankaerts, W., O'Sullivan, P., Burnett, A., & Straker, L. (2006). Altered patterns of superficial trunk muscle activation during sitting in nonspecific chronic low back pain patients: Importance of subclassification. *Spine, 31*, 2017–2023.
Davis, P. R. (1959). The medial inclination of the human thoracic intervertebral articular facets. *Journal of Anatomy, 93*, 68–74.
Drake, J. D. M., Aultman, C. D., McGill, S. M., & Callaghan, J. P. (2005). The influence of static axial torque in combined loading on intervertebral joint failure mechanics using a porcine model. *Clinical Biomechanics, 20*, 1038–1045.
Drake, J. D. M., & Callaghan, J. P. (2006). Elimination of electrocardiogram contamination from electromyogram signals: An evaluation of currently used removal techniques. *Journal of Electromyography and Kinesiology, 16*, 175–187.
Drake, J. D. M., & Callaghan, J. P. (2008). Do flexion/extension postures affect the *in vivo* passive lumbar spine response to applied axial twist moments? *Clinical Biomechanics, 23*, 510–519.
Drake, J. D. M., & Callaghan, J. P. (2009). Intervertebral neural foramina deformation due to two types of repetitive combined loading. *Clinical Biomechanics, 24*, 1–6.
Drake, J. D. M., Fischer, S. L., Brown, S. H., & Callaghan, J. P. (2006). Do exercise balls provide a training advantage for trunk extensor exercises? A biomechanical evaluation. *Journal of Manipulative and Physiological Therapeutics, 29*, 354–362.
Edmondston, S. J., Aggerholm, M., Elfving, S., Flores, N., Ng, C., Smith, R., et al (2007). Influence of posture on the range of axial rotation and coupled lateral flexion of the thoracic spine. *Journal of Manipulative and Physiological Therapeutics, 30*, 193–199.
Fleiss, J. L. (1986). *The design and analysis of clinical experiments*. New York: Wiley.
Gregersen, G. G., & Lucas, D. B. (1967). An in vivo study of the axial rotation of the human thoracolumbar spine. *The Journal of Bone & Joint Surgery, 49-A*, 247–262.
Kumar, S. (2010). EMG in rotation-flexion of the torso. *Journal of Electromyography and Kinesiology, 20*, 1146–1154.
Lee, H., & Granata, K. P. (2008). Process stationarity and reliability of trunk postural stability. *Clinical Biomechanics, 23*, 735–742.
Lee, P. J., Lee, E. L., & Hayes, W. C. (2013). The ratio of thoracic to lumbar compression force is posture dependent. *Ergonomics, 56*, 832–841.
Maiman, D. J., & Pintar, F. A. (1992). Anatomy and clinical biomechanics of the thoracic spine. *Clinical Neurosurgery, 38*, 296–324.
Marras, W. S., & Mirka, G. A. (1993). Electromyographic studies of the lumbar trunk musculature during the generation of low-level trunk acceleration. *Journal of Orthopaedic Research, 11*, 811–817.
McGill, S. M. (1991). Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: Implications for lumbar mechanics. *Journal of Orthopaedic Research, 9*, 91–103.
Montgomery, T., Boocock, M., & Hing, W. (2011). The effects of spinal posture and pelvic fixation on trunk rotation range of motion. *Clinical Biomechanics, 26*, 707–712.
Nairn, B. C., Chisholm, S. R., & Drake, J. D. M. (2013). What is slumped sitting? A kinematic and electromyographical evaluation. *Manual Therapy, 18*, 498–505.
Oxland, T. R., Lin, R. M., & Panjabi, M. M. (1992). Three-dimensional mechanical properties of the thoracolumbar junction. *Journal of Orthopaedic Research, 10*, 573–580.
Panjabi, M. M., Brand, R. A., & White, A. A. (1976). Mechanical properties of the human thoracic spine as shown by three-dimensional load-displacement curves. *The Journal of Bone & Joint Surgery, 58*, 642–652.
Pearcy, M. J., & Tibrewal, S. B. (1984). Axial rotation and lateral bending in the normal lumbar spine measured by three-dimensional radiography. *Spine, 9*, 582–587.
Preuss, R. A., & Popovic, M. R. (2010). Three-dimensional spine kinematics during multidirectional, target-directed trunk movement in sitting. *Journal of Electromyography and Kinesiology, 20*, 823–832.
Tortora, G. J. (2005). *Principles of human anatomy* (10th ed.). New York: Wiley.
Watkins, R., 4th., Watkins, R., 3rd., Williams, L., Ahlbrand, S., Garcia, R., Karamanian, A., et al (2005). Stability provided by the sternum and rib cage in the thoracic spine. *Spine, 30*, 1283–1286.
White, A. A., & Panjabi, M. M. (1990). *Clinical biomechanics of the spine* (2nd ed.). Philadelphia: JB Lippincott.
Willems, J. M., Jull, G. A., & Ng, J. K.-F. (1996). An in vivo study of the primary and coupled rotations of the thoracic spine. *Clinical Biomechanics, 11*, 311–316.
Williams, S. A. (2011). Evolution of the hominoid vertebral column. (Doctoral dissertation). Urbana, IL: University of Illinois at Urbana-Champaign.
Winter, D. A. (2009). *Biomechanics and motor control of human movement* (4th ed.). New York: Wiley.